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Feasibility of Using Piezohydraulic Pumps as Motors for Pediatric Ventricular Assist Devices*

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***Abstract*— The feasibility of using piezohydraulic pumps in drivers for pediatric ventricular assist devices is presented in this article. In this study a 0.5 kg piezohydraulic pump is incorporated into a ventricular assist device driver to drive a pulsatile pediatric 30 mL stroke ventricular assist device (VAD). The driver consists of a piezoelectric-hydraulic hybrid actuator and volume amplification section. Mechanical tests were performed on the pump and the hybrid actuator and a maximum power output of 5.4 W and 1.6 W were recorded respectively. The driver was tested running at multiple heart rates from 50-80 beats per minute (BPM) in an in-vitro bench top mock circulation to characterize the performance of the driver under a circulatory load. The maximum drive pressure output by the driver was 35 kPa. Peak flow rate from the VAD driven by the new driver was 6 L/min against a 10 kPa back pressure. Mean flow rate from the VAD outlet was 2.35 L/min for 80 BPM operation.**

I. INTRODUCTION

Cardiomyopathy, a disease in which the heart muscle functions abnormally, affects 1.13 in 100,000 children a year [1]. While pulsatile pediatric ventricular assist devices have provided short-term circulatory support for children with these diseases [2],[3], the drivers for these systems has limited patient mobility. Specifically, pulsatile VAD drivers rely on bulky electromagnetic motors and air compressors to sustain mechanical support in these small patients. Driver miniaturization is difficult due to the fact that the efficiencies and power output per unit volume of electromagnetic motors decreases as the physical size decreases [4]. Therefore, a new compact powerful motor is needed to further reduce the size of pulsatile VAD drivers. One approach for this problem is to utilize high frequency piezoelectric actuators.

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Piezoelectric frequency leveraged actuators have been used in standing and traveling wave ultrasonic motors [5] and inchworm motors [6]. These solid-state devices utilize a piezoelectric operated at high frequency to convert small strokes and low work output to large displacement and high power output [7]. Since power output per unit volume scales with size favorably in piezoelectric materials, these actuators can be miniaturized without sacrificing power density unlike electromagnetic motors. For example, Koc and Uchino [8] developed an 11 mm diameter rotary ultrasonic motor with a piezoelectric disc stator actuated at 110 kHz, producing 11 mW. Cagatay and Uchino also designed a 1.6 mm diameter ultrasonic motor with a power output of 45 mW [9]. In addition, Tieck *et al.* compared linear, horn-type, unidirectional rotary, and bidirectional traveling wave piezoelectric motors and found that the intrinsic piezoelectric material power densities ranged from 6.53 W/kg to 4.5 kW/kg while the actual power densities of the motors only spanned 0.82 W/kg to 85.73 W/kg [10]. All of these motor's relatively low power densities were attributed to the use of friction to transfer mechanical loads.

Piezohydraulic pumps, unlike ultrasonic and inchworm motors, do not utilize friction to transfer mechanical loads. These pumps consist of a piezoelectric actuators working with valves to generate hydraulic power in the form of a pressurized continuous flow [11]. For instance, Mauk and Lynch utilized a piezohydraulic pump to produce a device with a blocking force of 271 N and 72.5 mm/s no-load velocity (4.9 W maximum power output) [12]. However, the piezoelectric stack was operated at a relatively low

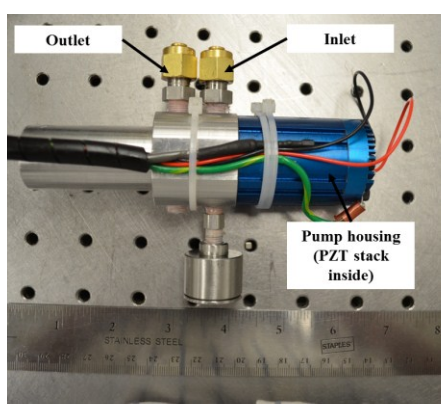

Figure 1. The 0.5 kg Kinetic Ceramics PHP3-1.6 used as the motor in the ventricular assist device driver

frequency (ie. < 100 Hz) limiting the power density. O'Neill and Burchfield produced piezohydraulic pumps operating in the kilohertz regime with power densities as high as 157 W/Kg and total power on the order of 46 W [13],[14]. While these pumps have produced large power outputs, the pump valve resonant frequency (<1 kHz) limits further increases in power output [15]. Future piezohydraulic pumps are anticipated to use these high frequency valves (~10 kHz) but they are not yet available. Nonetheless, it is important to begin characterizing piezohydraulic pumps for VAD motors.

The purpose of this research is to investigate the feasibility of using a frequency leveraged piezohydraulic pump (PHP) in a driver for pulsatile pediatric ventricular assist devices. A 0.5 kg PHP, designed for aerospace applications, is integrated into a driver consisting of a piezoelectric-hydraulic hybrid actuator (ie. hybrid actuator) and volume amplification section to drive a commercial pediatric VAD. The design and operating characteristics of the VAD driver are based on the cardiac cycle and the required power output for the heart. The driver is used to provide flow in a hydraulic mock circulation at the cardiac rhythm. Results indicate that a PHP is a feasible motor for pediatric ventricular assist devices.

II. EXPERIMENTAL SETUP

A. Piezohydraulic Ventricular Assist Device Driver

The piezohydraulic pump (Kinetic Ceramics; PHP3-1.6) used in this study is shown in Figure 1. The pump uses a 1.6" long PZT stack and metal diaphragm in the pump chamber to transfer the stored energy from the PZT to the hydraulic fluid. The piezoelectric and pump chamber are located in the blue anodized housing shown in Figure 1. Passive check valves in the inlet and outlet ports convert the mechanical stroke of the PZT to a continuous flow of hydraulic fluid. This device operates at 1.5 kHz with electric field oscillation from -0.5 to 2 MV/m. All tests were carried out with the piezohydraulic pump running at these settings.

The VAD driver used in this research, schematic shown in Figure 2a and picture shown in Figure 2b, consists of a piezoelectric-hydraulic hybrid actuator, referred to as the hybrid actuator, and a volume amplification section. The hybrid actuator consists of the PHP3-1.6, a 1.9 cm diameter double-rod hydraulic cylinder (Mack Corporation), a solenoid operated 4-way 2-position spool type directional valve (Hydraforce Inc.; SV08-40), and a 40 mL diaphragm-style hydraulic accumulator. The hybrid actuator uses Pennzoil Dexron VI as the working fluid which does not come into contact with blood. The hydraulic cylinder converts the fluidic output of the PHP into a linear mechanical stroke. The directional valve redirects the flow into the hydraulic cylinder producing bidirectional actuation. The hydraulic accumulator provides a bias pressure of 2.2 MPa to prevent cavitation and dampen large pressure changes in the system.

The volume amplification section consists of a 10 cm diameter pneumatic cylinder. The larger bore cylinder amplifies the stroke output while reducing the pressure produced by the piezohydraulic pump. A pneumatic driveline is used to connect the pneumatic cylinder to the pneumatic driveline connection on the VAD. The bidirectional motion of the pneumatic cylinder directly supplies pressurized air to the VAD during hydraulic cylinder rod extension or removes air from the VAD during rod retraction. For this study, the volume amplification section supplies air to a 30 mL stroke volume Berlin Heart EXCOR pulsatile VAD.

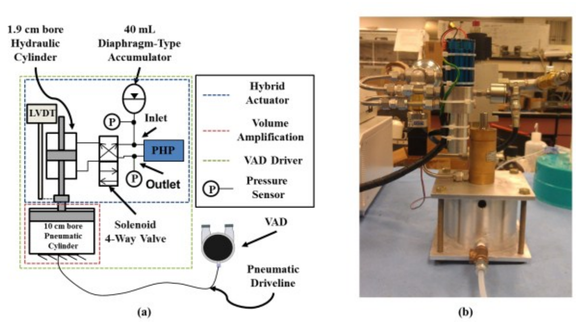

Figure 2. (a) schematic of the VAD driver consisting of a hybrid actuator and volume amplification section that supplies air to the pediatric VAD and (b) Picture of the assembled VAD driver

B. Mechanical Characterization

The hybrid actuator no-load velocity was measured by attaching a LVDT sensor (Omega Engineering; LD620) to the rod of the 19 mm diameter hydraulic cylinder. Pressures were also measured at the pump output and accumulator inlet with 4-20mA pressure transducers (Wika Instrument Corp.; A-10). Labview was used to collect the pressure and displacement data as well as to control the 4-way valve position for bidirectional hydraulic cylinder actuation (National Instruments; USB-6008, Labview 8.2).

A schematic of the hybrid actuator under an external mechanical load is shown in Figure 3. A 27 kN/m die spring was attached to the rod of the hydraulic cylinder for compression. Force output was calculated by measuring the compression of the spring. Velocity was calculated via the

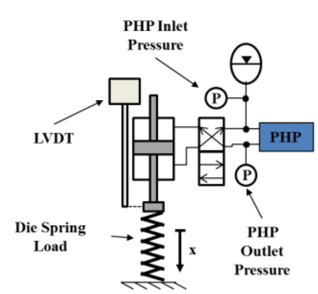

Figure 3. Schematic of the mechanical load test used to characterize power output of the PHP and the hybrid actuator

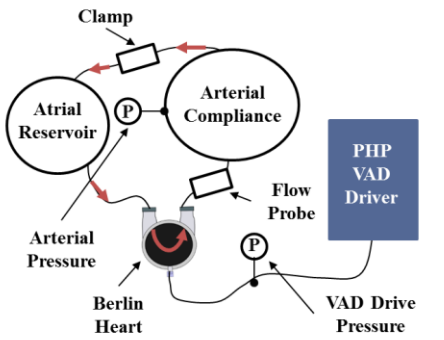

Figure 4. Schematic of the in-vitro mock circulation test setup used to characterize the performance of the VAD driver

rate of spring compression. PHP instantaneous power output was calculated by multiplying the flow rate measurements (ie. from velocity) and the pressure difference between the PHP inlet and outlet. In addition, hybrid actuator instantaneous power output was calculated from instantaneous force and velocity measurements (ie. Power = Force x velocity).

C. In Vitro Characterization

The *in-vitro* test used to quantify the performance of the VAD driver is shown in Figure 4. The VAD was inserted into a mock circulation consisting of an atrial reservoir, arterial compliance chamber (which served as the circulatory load), 3/8" ID Tygon tubing, and an ultrasonic flow probes and meter (Transonic Systems Inc., ME12PXL) to measure the flow rate. Room temperature glycerine and water mixture was used as the pumping medium in the mock circulation to mimic the viscosity of blood. The atrial reservoir and arterial compliance chambers are fabricated from 10 cm OD (9.5 cm ID) and 15.25 cm tall acrylic tubes [16]. The arterial chamber was closed to atmosphere. Arterial pressure and VAD driver pressure was recorded with medical pressure transducers (Utah Medical; Deltran).

During the *in-vitro* test, the hybrid actuator was controlled to mimic the two phase cardiac cycle, which consists of systole, ventricular blood ejection, and diastole, ventricular refilling. During systole, the 4-way hydraulic valve was turned off and the PHP was turned on to provide forward actuation of the hydraulic cylinder rod and driving pressure to the VAD. During diastole, the 4-way hydraulic valve was turned on and the hydraulic cylinder rod was allowed to retract to remove air from the VAD pneumatic chamber. The PHP was operated at a 95% duty cycle, turned on during systole and diastole, with a 25 millisecond off period between each phase. VAD drive pressure, PHP outlet pressure, PHP inlet pressure, and hydraulic cylinder rod displacement were also monitored.

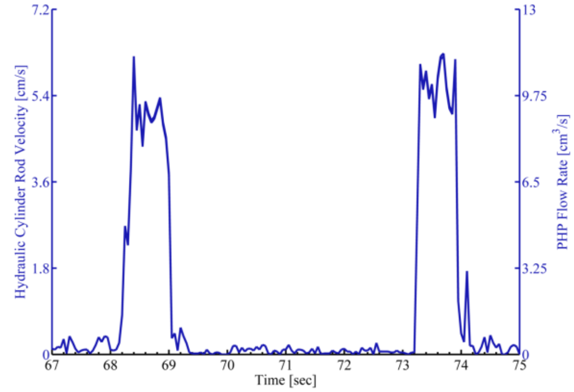

Figure 5. No load velocities (and PHP flow rate) for the hybrid actuator

III. RESULTS AND DISCUSSION

A. Mechanical Results

The 19 mm hydraulic cylinder rod velocity, left ordinate, versus time for both forward and reverse actuation in the absence of an external mechanical load is plotted in Figure 5. The maximum velocity is 5.7 cm/s, which corresponds to a PHP output flow rate of 10.2 cm^3/s (right ordinate). It is important to note that the maximum velocity of the actuator can be increased by decreasing the hydraulic cylinder diameter. Since the flow requirements for typical pediatric VADs ranges between 1-5 L/min, the flow from the output of a newly designed pump would need to be an order of magnitude higher.

The change in pressure across the piezohydraulic pump (left ordinate) versus the PHP flow rate is plotted in Figure 6. The data was fitted using a least squares fitting approach. As the load is increased, the pump flow rate decreases linearly with increasing back pressure. The pump power output, the product of pressure difference and flow rate, increases parabolically with increasing flow rate as seen in

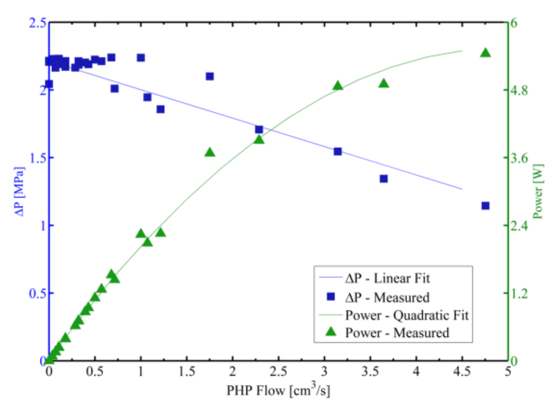

Figure 6. PHP outlet and inlet Pressure difference (left ordinate) versus flow rate and PHP power output (right ordinate) versus flow rate in the presence of an external load

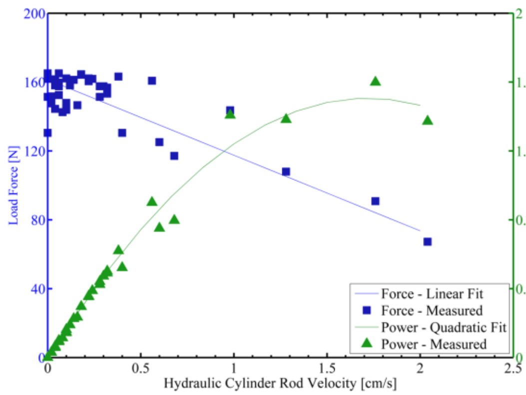

Figure 7. Hybrid actuator output force (left ordinate) versus velocity and hybrid actuator power output (right ordinate) versus velocity in the presence of an external load

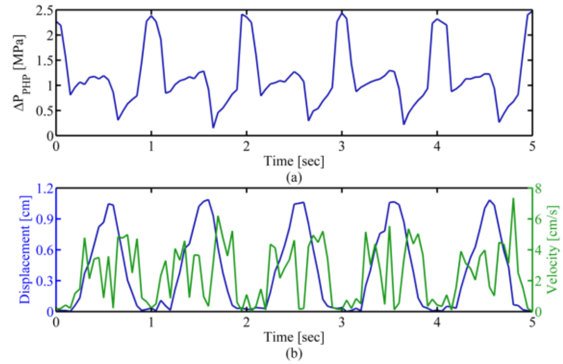

Figure 8. (a) . PHP outlet and inlet pressure difference and (b) hydraulic cylinder displacement (left ordinate) and hydraulic cylinder rod velocity (right ordinate)

the right ordinate. The peak pump power output measured is 5.4 W at a 5 cm^3/s flow rate. Typically, this pump can output up to 28 W of power [17]. Nonetheless, the 5.4 W power output exceeds the power of the normal pediatric left ventricle (1 W), which suggests that a well-designed pump should be five times smaller than tested in this study. If the pump were operated at a higher frequency (10 kHz range), the pump size could theoretically be scaled down further (ie. fifty times).

The load force, left ordinate, as a function of hydraulic cylinder rod velocity is shown in Figure 7. Force and velocity data were fitted using least squares fitting. The shaft velocity of the hybrid actuator decreases linearly as the applied external load increases. The blocking force is 180 N. The right ordinate shows the power output of the overall hybrid actuator (ie. the product of spring force and velocity) as a function of velocity. The maximum power output in the forward direction is 1.6 W at a 1.75 cm/s shaft velocity. This indicates that the actuator should supply sufficient force and velocity to the volume amplification section to ensure full ejection from the VAD.

B. *In-vitro Results*

The pressure difference across the PHP in a five second interval during the 60 BPM in-vitro test are shown in Figure 8a. The PHP inlet pressure (ie. bias pressure) stays constant at 2.2 MPa while the piezohydraulic pump outlet pressure periodically increases to actuate the pneumatic cylinder. Peak PHP outlet pressure is caused by the pressure buildup when the hydraulic cylinder is fully retracted during each cycle. Hydraulic cylinder rod displacement and velocity are plotted in Figure 8b. The maximum rod stroke length under the 60 BPM control is 1 cm and the magnitude of the forward and backward velocity was 4 cm/s. This shows that the actuator operates at a velocity close to the no-load velocity of the hybrid actuator (5.7 cm/s) when subjected to a circulatory load. This corresponds to a small force output to the VAD and an impedance mismatched system. To correct this, a newly designed pump with the appropriate pressure output and flow rate needs to be developed.

Ventricular assist device drive pressure and arterial pressure (left ordinate) and VAD flow output (right ordinate) are shown in Figure 9. The peak VAD drive pressure was 35 kPa, which compares well to commercial pulsatile VAD drivers which output between 25-40 kPa. A peak flow rate of 6 L/min against a 10 kPa arterial pressure was recorded. This corresponds to a total of 1 W imparted to the mock circulation from the current drive setup Traditional drivers can output between 0-10 L/min instantaneous flows against arterial pressures ranging from 10-16 kPa. Therefore, the initial driver prototype we have proposed confirms that it is feasible for piezohydraulic pumps to output pressures and flow rates that are comparable to traditional pulsatile VAD drivers.

Figure 10 shows the mean flow rate from the VAD versus heart rate for the in-vitro experiment when the VAD driver was operated at different heart rates. The result is compared to the target flow. Like traditional pulsatile VAD drivers, the piezohydraulic actuated driver outputs sufficient pressure to the VAD to achieve a mean flow that is linearly related to the

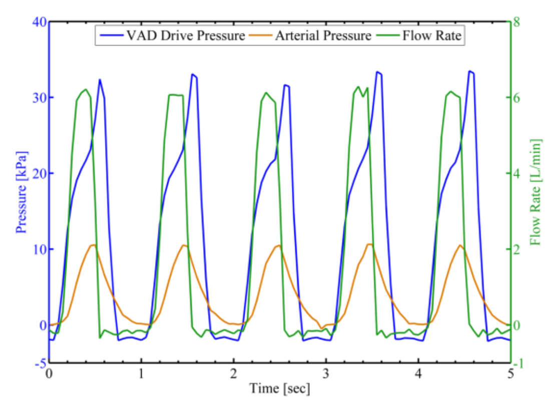

Figure 9. VAD drive pressure, arterial pressure (left ordinate), and flow rate (right ordinate) wave forms during the 60 BPM in-vitro test

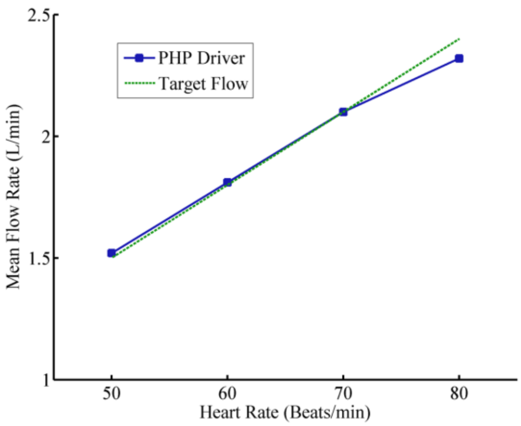

Figure 10. Mean flow from the piezohydraulic driven VAD compared with a traditional driver

heart rate below 80 BPM. Thus for heart rates ranging from 50-80 BPM, the full 30 mL stroke of the VAD was ejected to the circulation. Heart rates of 150 BPM with mean flow of 4.5 L/min can be achieved by increasing the volume amplification such that sufficient volume is delivered to the VAD for small displacements during higher heart rate operations. Nonetheless, even with a larger volume amplification section, the entire system can be contained in a 15 x 15 x 30 cm volume. This is much smaller than the stationary drive units that are typically 46 x 95 x 73 cm in size.

IV. Conclusion

The feasibility of using a piezohydraulic actuated ventricular assist device driver has been demonstrated. Mechanical tests were performed to characterize the maximum velocities of the hybrid actuator under no external load. Spring load tests were also performed to characterize the maximum power output the hybrid actuator could transfer to an external load. For this actuator, 5.7 cm/s velocities were recorded. In addition, a maximum PHP power output of 5.4 W was recorded and is higher than the 1 W that is typically output by the healthy pediatric heart. The hybrid actuator and volume amplification section were incorporated as a VAD driver and tested *in-vitro*. At 60 BPM, the driver provided sufficient VAD drive pressure to maintain a 1.8 L/min flow rate with a 35 kPa drive pressure. The driver was also able to operate at 80 BPM outputting 2.35 L/min of mean flow. These results indicate that a smaller pump can be used to drive a VAD directly when losses are reduced, piezoelectric actuation frequency is increased, and the pump is developed for high flow and low pressure requirements, which translates to a smaller driver that will allow patients to recover quickly.

References

[1] S. E. Lipshultz, L.A. Sleeper, Towbin, J. A., Lowe, A. M., Orav, E. J., Cox, G. F., Lurie, P. R., McCoy, K. L., McDonald, M. A., Messere, J. E., and Colan, S. D. 2003 "The incidence of pediatric cardiomyopathy in two regions of the United States," *New England Journal of Medicine*, 348(17):1647–1655.

[2] F. Fynn-Thompson and C. Almond, "Pediatric Ventricular Assist Devices," *Pediatr Cardiol*, vol. 28, no. 2, pp. 149–155, Jan. 2007.

[3] A. L. Throckmorton and S. G. Chopski, "Pediatric Circulatory Support: Current Strategies and Future Directions. Biventricular and Univentricular Mechanical Assistance," *ASAIO Journal*, vol. 54, no. 5, pp. 491–497, Sep. 2008.

[4] Uchino, S. Cagatay, B. Koc, S. Dong, P. Bouchilloux, and M. Strauss, "Micro Piezoelectric Ultrasonic Motors," *Journal of Electroceramics*, vol. 13, no. 1–3, pp. 393–401, 2004.

[5] K. Uchino, "Piezoelectric ultrasonic motors: overview," *Smart Materials and Structures,* vol. 7, no. 3, pp. 273–285, 1998.

[6] T. Galante, J. Frank, J. Bernard, W. Chen, G. A. Lesieutre, and G. H. Koopmann, "Design, Modeling, and Performance of a High Force Piezoelectric Inchworm Motor," *Journal of Intelligent Material Systems and Structures*, vol. 10, no. 12, pp. 962–972, Dec. 1999.

[7] D. J. Leo, , "Motion Control Appications," in *Engineering Analysis of Smart Material Systems* Hoboken, NJ: John Wiley & Sons, ch. 8, sec. 8.3, pp. 402-409, 2007.

[8] B. Koc, A. Dogan, Y. Xu, R. E. Newnham, and K. Uchino, "An Ultrasonic Motor Using a Metal-Ceramic Composite Actuator Generating Torsional Displacement," *Japanese Journal of Applied Physics*, vol. 37, no. Part 1, No. 10, pp. 5659–5662, Oct. 1998.

[9] S. Cagatay, B. Koc, and K. Uchino, "A 1.6-mm, metal tube ultrasonic motor," *IEEE Transactions on Ultrasonics, Ferroelectrics and Frequency Control*, vol. 50, no. 7, pp. 782–786, Jul. 2003.

[10] R. M. Tieck, M. C. Emmons, and G. P. Carman, "Alternate Evaluation Criteria of Piezoelectric Motors," *Journal of Intelligent Material Systems and Structures*, vol. 18, no. 11, pp. 1215–1221, Nov. 2007.

[11] N. W. Hagood, D. C. Roberts, L. Saggere, K. S. Breuer, K.-S. Chen, J. A. Carretero, H. Li, R. Mlcak, S. W. Pulitzer, M. A. Schmidt, S. M. Spearing, and Y.-H. Su, "Microhydraulic transducer technology for actuation and power generation," *Proceedings of SPIE*, vol. 3985, no. 1, pp. 680–688, Jun. 2000.

[12] L. D. Mauck and C. S. Lynch, "Piezoelectric Hydraulic Pump Development," *Journal of Intelligent Material Systems and Structures*, vol. 11, no. 10, pp. 758–764, Oct. 2000.

[13] C. O'Neill and J. Burchfield, "Kinetic ceramics piezoelectric hydraulic pumps," presented at the Industrial and Commercial Applications of Smart Structures Technologies 2007, San Diego, California, USA, 2007, p. 65270I–65270I–14.

[14] R. M. Tieck, "Characterization of a piezohydraulic actuator," presented at the Smart Structures and Materials 2005: Smart Structures and Integrated Systems, San Diego, CA, USA, 2005, pp. 671–679.

[15] D. G. Lee, D. D. Shin, and G. P. Carman, "Large flow rate/high frequency microvalve array for high performance actuators," *Sensors & Actuators: A. Physical*, vol. 134, no. 1, pp. 257–263, 2007.

[16] S. Vandenberghe, F. Shu, D. K. Arnold, and J. F. Antaki, "A simple, economical, and effective portable paediatric mock circulatory system," *Proc Inst Mech Eng H*, vol. 225, no. 7, pp. 648–656, Jul. 2011.

[17] C. O'Neill, Kinetic Ceramics Inc., Personal discussion, 2012.